The Flow-Pressure Plot: A New Look on the Patient-Ventilator Interaction in Neonatal Care

Hugo Devlieger, Thierry Bayet,† Jacques Lombet,‡ Stéphane Naudé,* and Christian Eugène†*

Most modern neonatal ventilators have now a built-in flow sensor and, as a spin-off of their mechanical action, provide some information about lung function characteristics as compliance and resistance after computation of the flow and pressure signals. Additionally, respiratory graphics as volume-pressure and flow-volume plots can be displayed. In clinical practice, however, they are rarely used to refine the ventilator setting. A nonconventional flow-pressure plot is presented in this article, constructed from the volume or flow, and pressure outputs of a Babylog 8000 (Dräger, Lübeck, Germany). This flow-pressure diagram appears to be useful in the real-time computation of respiratory mechanics based on the Rahn's law of respiratory motion. Its major advantage, however, is the easy pattern recognition of subtle changes in infant-ventilator interaction, ie, excessive triggering, fighting against the ventilator, augmented breath, tube subobstruction. It may be useful to add the flow-pressure plot to the classical respiratory graphics allowing to monitor mechanical ventilation more accurately and to fine tune the ventilator setting accordingly.
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Mechanical ventilation is since long a standard practice in neonatal intensive care, which has largely contributed to improve survival and quality of survival of the very low birth weight infants. However, most practice is based on consolidated clinical experience, rather than on knowledge of the conditions leading to respiratory failure and of the response of the respiratory system to the therapy. Each center set up guidelines that are considered to be the best for all patients.[1] For many years indeed, it was impossible to evaluate the respiratory mechanics otherwise than by watching the chest wall movements and by monitoring blood gases. Techniques of lung function assessment have been adapted since and moved from the laboratory to the neonatal intensive care unit. As a result, an increasing number of lung function tests and techniques have been proposed to assess the severity of lung disease and to refine management.[2-11]

The usefulness of these techniques has been questioned.[12] Indeed, to be applicable in clinical practice these techniques need to fulfill the following prerequisites: 1) They should be in line with the minimal handling principle that governs most neonatal intensive care units; and 2) Disconnection of the ventilator should be avoided and dead space or additional resistance should be limited. To be in line with these requirements, a number of manufacturers have proposed pressure limited time-cycled ventilators with a built-in lung function assessment program based on a integrated flow sensor signal and pressure measurement. However, some questions remain for the practising neonatologist, eg, has lung function assessment some diagnostic and prognostic added value?[13] Can lung function assessment help to refine the ventilator setting? Can lung function inform the clinician about the patient-ventilator interaction and hence conduct the sedation policy?

This study investigates how the data generated by a neonatal ventilator with a sensitive flow

*From the *Department of Pediatrics, Katholieke Universiteit Leuven; the ‡Department of Pediatrics, State University of Liège; and the †Departement of Electricity, Laboratory of Electrotechnics and Instrumentation, Université Catholique de Louvain.*
Address reprints requests to Hugo Devlieger MD, PhD, Department of Pediatrics, University Hospital Gasthuisberg, Herestraat 49, B-3000 Leuven, Belgium; e-mail: hugo.devlieger@uz.kuleuven.ac.be.
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sensor can be computed to yield on-line useful information on the respiratory mechanics of the infant lung. In addition to the classical pressure-volume and flow-volume plots, a nonconventional flow-pressure plot is presented that appeared to be useful in both the computation of respiratory mechanics and in the identification of subtle changes in the interaction between the newborn infant and the ventilator.

System Description

The ventilator (Babylog 8000; Dräger, Lübeck, Germany) is a continuous flow, time-cycled, pressure limited ventilator with an integrated flow sensor based on a heated wire anemometer. The ventilator has 2 programmable analogue outputs of which pressure (−10 to +90 mBars) and volume (0-100 mL) were selected (Fig 1). These analogue signals are relayed to an A/D converter. The acquisition card (PC-30-C; Boston Technology, England) is supported not only for the purposes of single A/D conversions into 12-bit digital codes, but support all forms of sampling, including both interrupt driven sampling and direct memory access driven sampling. The direct memory access controller limits the acquisition to 32K octets per measurement. This results in an acquisition limited to 20 seconds for a sampling rate of 80 Hertz that is necessary to accurately reproduce the signals.

The configuration of the PC-30 board and of the PC is supported by a software program written in Turbo C to yield a number of user-friendly functions as the time frame for acquisition and computation of the signals, the selection of the signals and plots to be visualized, details about the graphical basis of computation and of the statistics used, and a saving function of an given set of data. A time window of 10 seconds was mostly used because of the limitations of the acquisition card.

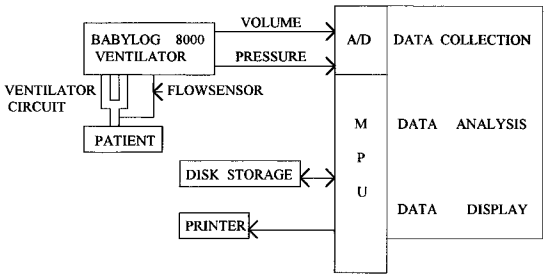

Figure 1. Block diagram illustrating the system. The newborn infant is ventilated with a time-cycled pressure limited ventilator with an integrated flow-sensor. The volume and pressure analogue signals are relayed to an A/D converter, introduced in a computer, and transformed and arranged into useful numerical and graphical outputs.

The Output

Apart from the raw signals of the volume, flow and pressure versus time, classical volume-pressure and flow-volume plots were computed and could be visualized on the screen. In addition, a flow-pressure plot (Fig 2) was computed. Although not a classical plot in respiratory graphics, this plot has been added because it appeared to yield useful information, allowing to calculate the inspiratory and expiratory resistance as well as the dynamic compliance. In passively ventilated patients, the plot consists of 3 distinctive lines connected by 2 transients. The line A is a rapidly raising straight line starting at flow 0 and pressure equal to end expiratory pressure (EEP). This delta flow versus delta pressure is indicative of the inspiratory resistance. The straight slope lasts about 50 ms. Next, a nearly straight horizontal line is drawn indicative of a maximum or threshold flow. At the condition that the change in pressure per unit of time applied by the ventilator remains a constant during maximal inspiratory flow, this straight horizontal line represents the dynamic compliance as measured in the steepest inspiratory part of the volume pressure loop or represents the dynamic compliance as measured by the least mean square analysis method. Indeed ($flow_{max}$/dp/dt) = (dv/dt at threshold/dp/dt) = (dv at threshold/dp) = Cdyn. After a transient which represent the flow fall to zero, a nearly straight line C is drawn which represents the quasi linear relationship between pressure fall and flow fall in the early expiration and is indicative of early expiratory resistance. During the rest of expiration, flows fall further while systemic pressure remains at the EEP level.

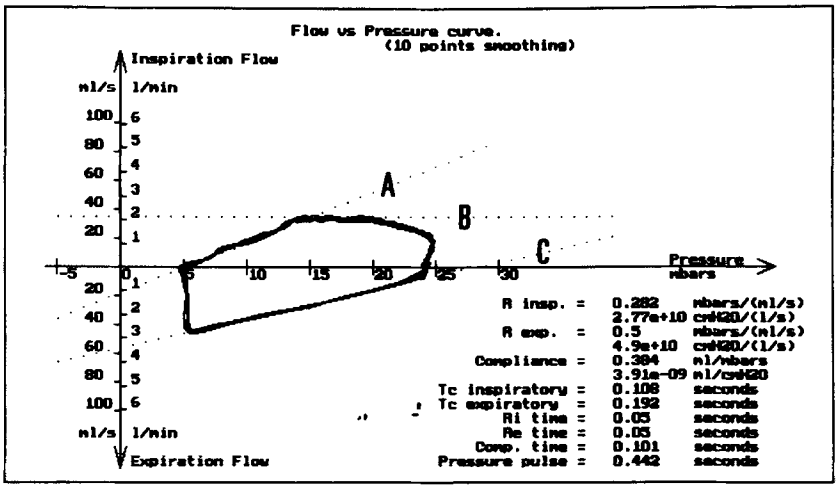

Figure 2. The flow-pressure plot allows calculating the inspiratory and expiratory resistance and the dynamic compliance. Line A starting at zero flow and at a pressure equal to the end-expiratory pressure, represents Δflow/ΔP and is indicative of inspiratory resistance. Line B, the nearly straight horizontal line, indicates maximum flow and is representative of dynamic compliance. Line C represents the linear relationship between the alveolar to circuit pressure gradient and pressure and is indicative of resistance in the early expiratory phase (see text for further explanation). The calculated resistance, compliance, and time constants are displayed.

Physiological Basis of the Flow/Pressure Plot

Rahn's law of respiratory motion is the basis of the calculation of the respiratory indices. Indeed this law splits the driving pressure of respiratory motion up into three parts: = (R × flow) + (1/C × volume) + (I × turbulent flow) in which P is the pressure applied, R is the resistance, C is the compliance and I the inertance. Because in conventional ventilation, flow rate is kept low, the last part of the equation can be skipped and the equation can be reduced to P = (R × flow) + (1/C × volume).

Figure 3 gives a theoretical and schematic representation of the change in pressure gradient between the pressure in the ventilator circuit applied to the airways, and the intra-alveolar pressure. This pressure gradient is the driving force of the flow entering and leaving the alveoli and represented in the flow-pressure plot. The consistency of the lines A, B, and C can be explained by the pressure gradients induced consecutively in the respiratory system between the ventilator circuit and the alveoli. Indeed, during early inspiration a short but increasing gradient of pressure is generated during which the inspiratory resistance is measured. This short (50 ms) build-up of a pressure gradient is followed by a nearly constant gradient between airway opening pressure and alveolar pressure in the ascending part of the pressure versus time loop during which the dynamic compliance is measured. During expiration, the short period (50 ms) during which the circuit pressure falls to EEP, gives rise to a pressure gradient between

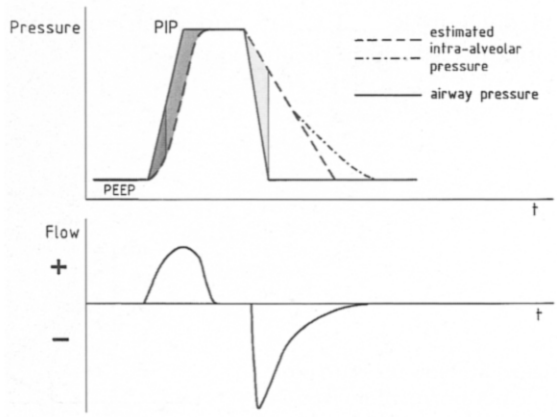

Figure 3. Schematic representation of the variation of the airway- and estimated intra-alveolar pressure over time in combination with the flow signal. The grey areas correspond respectively to the line A, B, and C in Figure 2 and represent the epochs at which the calculations of compliance and resistance are computed.

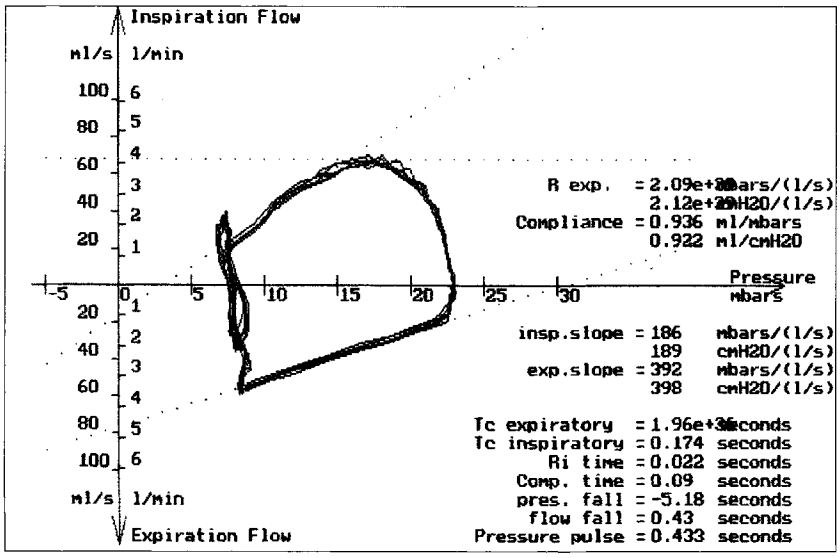

Figure 4. Flow-pressure plot of a newborn infant ventilated with a SIMV modus who triggers each ventilator beat in addition to his own respiratory activity.

the systemic pressure and the pressure in the alveoli which induces a steadily increasing expiratory flow from which the expiratory resistance can be calculated. The software program calculates the regression coefficient for the line A and C indicative of the in- and expiratory resistance. To calculate R_{exp}, the assumption is made that the alveolar pressure fall is linear over time. As a consequence, the alveolar pressure at the time EEP is reached in the circuit represents the pressure gradient at maximal expiratory flow allowing calculating R_{exp}. In addition to the numerical data on R and C, in- and expiratory time constants are numerically expressed.

Application in the Neonatal Intensive Care Unit

The method described allows the on-line assessment of lung function without disconnecting the infant from the ventilator and without augmenting the dead space. Apart from information about lung function, the flow-pressure plot draws a clear view on the interaction between the patient and his ventilator. Active inspiration and triggering can easily be identified (Fig 4) as well as strength the inspiratory effort (Fig 5). Fighting against the ventilator is equally well recognized (Fig 6). Subtle findings as the in-

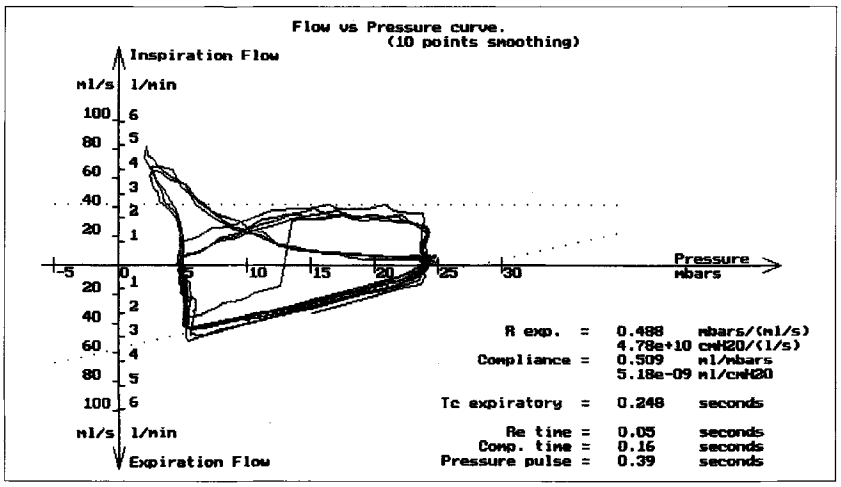

Figure 5. Flow-pressure plot of a newborn infant ventilated with a SIPPV modus who makes strong respiratory efforts. During the 10-second period, however, passive ventilation occurred 4 times while fighting against the ventilator was observed in one respiratory cycle.

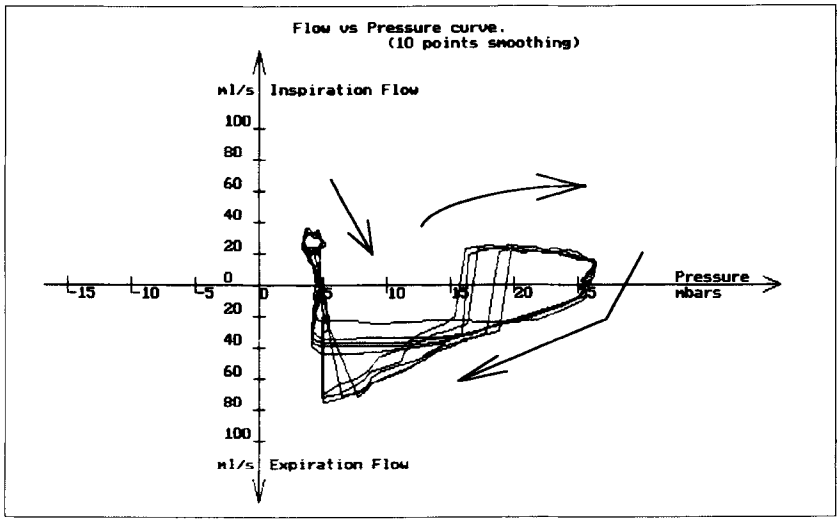

Figure 6. Flow-pressure plot of an infant who was permanently fighting against the ventilator when on SIMV.

spiratory augmenting reflex (Head's reflex) as a result of inflation can be identified as a steep flow increase (Fig 7). Some degree of hyperinflation can be visualized by a trend of the line B toward the x-axis instead of running parallel to the x-axis, indicating a fall of flow as a result of a decrease in pressure gradient.

Validation of the Compliance and Resistance Measurements

In 18 preterm ventilated infants measurement of compliance and resistance was performed simultaneously by the classical Sensor Medic PT2600 lung function device (Sensor Medic, Yorba Linda, CA) and by the method described above based on calculation of the regression coefficient of line B, A and C. For each infant, both the dynamic compliance (C_{dyn}) and the compliance of the respiratory system (C_{RS}) by means of a single breath occlusion technique, were determined with the Sensor Medic device. The results were compared by linear regression and a coefficient of correlation was calculated. Figures 8 and 9 show the results of the comparison between the compliance calculated from the flow-pressure plot (C_{dyn} B8000) and the C_{dyn} and C_{RS} measured with the Sensor Medic. There was a

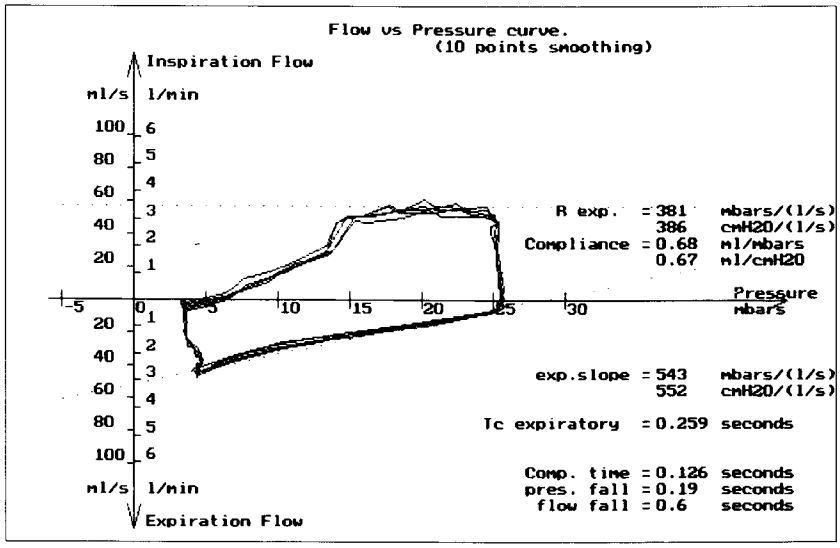

Figure 7. Flow-pressure plot of an infant with a inspiratory augmenting reflex (Head's reflex on each ventilation).

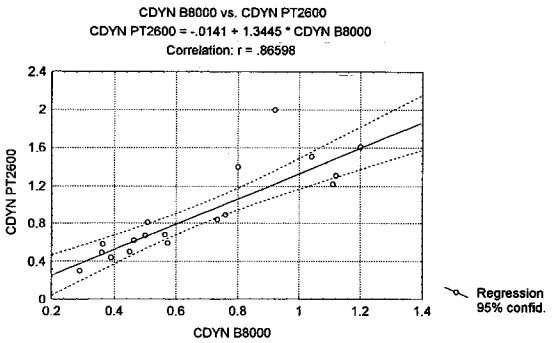

Figure 8. Comparison between dynamic compliance calculated from the flow-pressure plot (CDYN B8000) and C_{dyn} measured with the Sensor Medic (CDYN PT2600).

good correlation ($P < .05$) on linear regression both with the C_{dyn} and C_{RS} measured with the Sensor Medic, the correlation coefficient being 0.8. At high compliance values, however, there is an underestimation as compared to the dynamic compliance but not when measured as C_{RS}. It results that the compliance measured by way of the flow-pressure plot correlates best with the C_{RS}. Linear regression of the resistance measured by both methods showed a significant correlation ($P < .05$) with a correlation coefficient of 0.7 but a systematic underestimation occurred with lower values by the Sensor Medic technique.

General Discussion and Conclusion

One should keep in mind that the techniques for measuring respiratory mechanics are based on a linear single compartment model. It assumes that the respiratory system is reduced to a single balloon and a single airway with laminar airflow, and that the respiratory muscles are inactive. These conditions are seldom encountered during mechanical ventilation, especially with lung disease. As a result, these techniques are invalidated when comparing different patients but the information obtained is still relevant for patient monitoring and ventilator adaptation.

The system, described here, allows on-line, both numerical and graphical representation of respiratory characteristics in ventilated newborn infants. Limitations of the system result mainly from the sensitivity and frequency response of the flow sensor and pressure measurement. In addition, the characteristics of the ventilator, the circuit and the endotracheal tube play a crucial role in the computation of the respiratory function characteristics. Only part of them can be ascribed to the infant. This is particularly true for the measurement of resistance. As with many other methods of lung function described in the literature, the resistance is highly dependent on the resistance characteristics of the endotracheal tube.[14,15] However, as long as the ventilator, the circuit and the endotracheal tube remain unchanged including the degree of tube leakage, all variability in results stems from changes in the respiratory condition of the infant. In addition, when applied to adjust or modify the ventilator setting, a more pragmatic approach of the respiratory characteristics of the whole system including ventilator, circuit, E-T tube, and lung and chest wall of the infant and not just those of the patient's lung, is mandatory.

Despite these limitations, the on-line measurements of compliance and resistance and the visualisation of the volume-pressure, the flow-volume and especially the flow-pressure loop may help the clinician to identify subtle maladaptation between patient and ventilator and to correct it. Indeed, the major spin-off of this non-conventional F-P loop, described here, is the pattern recognition of subtle changes in infant-ventilator interaction such as excessive triggering, fighting against the ventilator, augmented breath, and tube sub-obstruction. In conclusion,

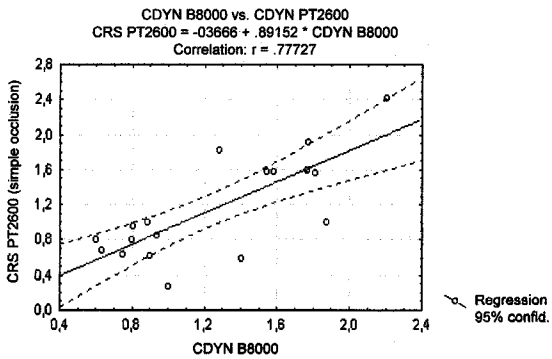

Figure 9. Comparison between the dynamic compliance calculated from the flow-pressure plot (CDyN B8000) and C_{RS} measured with the Sensor Medic (CRS PT1200).

in ventilated infants, the flow-pressure plot may offer definite advantages in the noninvasive on-line assessment of the respiratory system and can be added advantageously to the classical respiratory graphics in neonatal ventilators.

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